

(12) INTERNATIONAL APPLICATION PUBLISHED UNDER THE PATENT COOPERATION TREATY (PCT)

(19) World Intellectual Property Organization  
International Bureau



(43) International Publication Date  
30 November 2000 (30.11.2000)

PCT

(10) International Publication Number  
**WO 00/72050 A1**

- (51) International Patent Classification<sup>7</sup>: **G01V 3/00**
- (21) International Application Number: **PCT/US00/14171**
- (22) International Filing Date: **23 May 2000 (23.05.2000)**
- (25) Filing Language: **English**
- (26) Publication Language: **English**
- (30) Priority Data:  
**60/135,645** **24 May 1999 (24.05.1999)** **US**
- (71) Applicants and  
(72) Inventors: **KYRIAKOS, Walid, E. [LB/US];** 7 Euston Street, Apt. #4, Brookline, MA 02446 (US). **KACHER, Daniel, F. [US/US];** 58 Ridgemont Street, Apt. #2, Allston, MA 02134 (US).

(81) Designated States (*national*): AE, AG, AL, AM, AT, AU, AZ, BA, BB, BG, BR, BY, CA, CH, CN, CR, CU, CZ, DE, DK, DM, DZ, EE, ES, FI, GB, GD, GE, GH, GM, HR, HU, ID, IL, IN, IS, JP, KE, KG, KP, KR, KZ, LC, LK, LR, LS, LT, LU, LV, MA, MD, MG, MK, MN, MW, MX, MZ, NO, NZ, PL, PT, RO, RU, SD, SE, SG, SI, SK, SL, TJ, TM, TR, TT, TZ, UA, UG, US, UZ, VN, YU, ZA, ZW.

(84) Designated States (*regional*): ARIPO patent (GH, GM, KE, LS, MW, MZ, SD, SL, SZ, TZ, UG, ZW), Eurasian patent (AM, AZ, BY, KG, KZ, MD, RU, TJ, TM), European patent (AT, BE, CH, CY, DE, DK, ES, FI, FR, GB, GR, IE, IT, LU, MC, NL, PT, SE), OAPI patent (BF, BJ, CF, CG, CI, CM, GA, GN, GW, ML, MR, NE, SN, TD, TG).

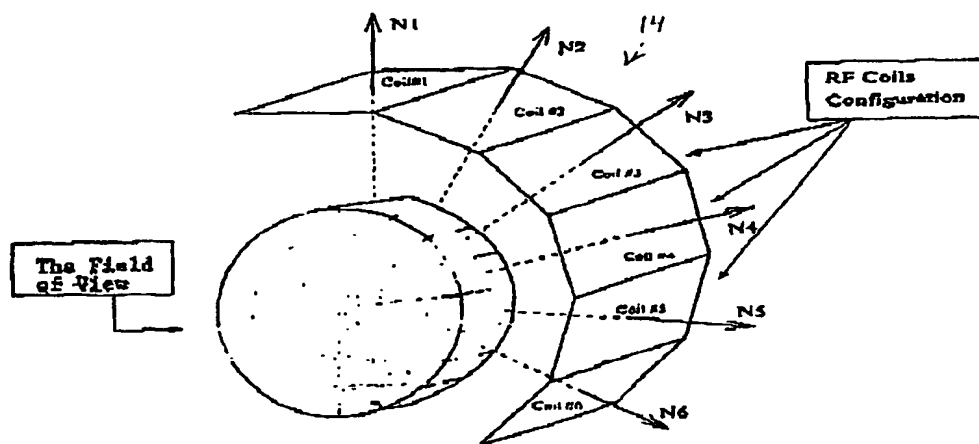
**Published:**

— *With international search report.*

- (74) Agent: **JOAO, Raymond, A.;** Ruskin, Moscou, Evans & Faltischek, P.C., 170 Old Country Road, Mineola, NY 11501-4366 (US).

*For two-letter codes and other abbreviations, refer to the "Guidance Notes on Codes and Abbreviations" appearing at the beginning of each regular issue of the PCT Gazette.*

(54) Title: **METHOD AND APPARATUS FOR PARALLEL DATA ACQUISITION FROM A MRI COIL ARRAY**



Schematic representation in 3-D of the Field of View and the RF pickup coil array positioning in this method. The coil orientations are described by vectors orthogonal to the coils.

- (57) Abstract: Magnetic resonance data is acquired in parallel by an array of separate RF coils (2, 4, 6) arranged circumferentially around the imaging subject. The apparatus and method operate on the basis of determining an estimate of the sensitivity profile of each coil in the array.

WO 00/72050 A1

## METHOD AND APPARATUS FOR PARALLEL DATA ACQUISITION FROM A MRI COIL ARRAY

BACKGROUND

## 5           Field of Invention

The present invention generally relates to apparatus and methods for nuclear magnetic resonance imaging (NMRI), also known as magnetic resonance imaging (MRI). More specifically, the invention relates to apparatus and methods for decreasing  
10   magnetic resonance (MR) data acquisition times, wherein magnetic resonance data is acquired in parallel using an array of receiver coils at least partially surrounding the object of interest, and the desired MRI image is then reconstructed in parallel.

## 15           Summary of the Prior Art

In dynamic MRI applications, such as functional imaging, interventional imaging and cardiac imaging, there is has long been a need in the art for methods and apparatus that provide high quality (e.g., high-resolution and signal-to-noise ratio)  
20   images. Conventional MRI imaging apparatus and methods, however, operate at speeds that are an order of magnitude slower than those which are currently deemed to be desirable. Some of these conventional methods are described in the Background section of U.S. Patent No. 5,365,172 to Hrovat et  
25   al. for "Methods and Apparatus for MRI", the disclosure of which is incorporated herein by reference.

In an attempt to attain faster operating speeds, several so-called "parallel" encoding apparatus combinations and/or methods have been developed. These apparatus combinations  
30   and/or methods rely on the use of multiple receiver coils for the acquisition of magnetic resonance data and high-speed data processors for the reconstruction of the field of view with significantly smaller data sets.

Among the parallel imaging techniques described in the literature, the work of Kwiat et al. ("A Decoupled Coil Detector Array for Fast Image Acquisition in Magnetic Resonance Imaging", *Medical Physics*, 18:251, 1991, the disclosure of which is hereby incorporated herein by reference) is significant. This work involved the investigation of methods for solving the inverse source problem on magnetic resonance signals received in multiple RF receiver coils. The technique proposed required the use of a number of RF coils equal to the number of pixels in the desired image. It also required that the sensitivity of the coils used be increased by an order of magnitude. Since these requirements are quite impractical in conventional magnetic resonance imaging (wherein the usual number of pixels in the image is on the order of 256x256), this technique has never been used successfully in a biological imaging experiment.

Other so-called "parallel" imaging techniques that use one dimensional sensitivity profiles of RF coils to encode space in a MRI context also have been proposed. For example, Ra, et al. ("Fast Imaging Using Sub-encoding Data Sets From Multiple Detectors", *Magn. Reson. Med.*, 30:142, 1993, the disclosure of which is hereby incorporated by reference herein), describes a method that uses sets of equally spaced k-space lines from multiple receiver coils in a line, and combines them with the one dimensional sensitivity profile information to remove the aliasing that occurs due to undersampling. A four-fold decrease in the image acquisition time of a water phantom was postulated to be possible by using an array of four coils.

Nevertheless, no biological images were shown in this article. It is believed that this may be indicative of a possible lack of robustness of the aliasing removal algorithm in practical situations.

A method called SMASH proposed by Sodickson et al. ("Simultaneous Acquisition Of Spatial Harmonics (SMASH): Fast Imaging With Radio Frequency Coil Arrays", *Magn. Reson. Med.*, 38:591-603, 1997, the disclosure of which is hereby  
5 incorporated herein by reference) has been found to be practical, and to yielded good results. SMASH enhances imaging speed by using multiple RF receiver coils. More specifically, it uses linear combinations of the 1D sensitivity profiles of receiver coils (weighted so as to form  
10 sinusoidal harmonics) of a one dimensional array to generate all k-space lines from a small subset of collected magnetic resonance data.

The SMASH method, however, is somewhat limited. It has an inherent inflexibility in the choice of the imaging plane  
15 to be viewed. Also, it has a demonstrable limitation in depth penetration. Further, there is a practical, physical limit on the number of coils that can be placed along one direction in a magnetic resonance imaging apparatus - particularly if the coils are to be de-coupled from one another.

#### SUMMARY OF THE INVENTION

The present invention provides an apparatus and a method that significantly decrease both magnetic resonance data  
25 acquisition time, and image reconstruction time, in magnetic resonance imaging.

Also, the present invention provides an apparatus and a method wherein sets of magnetic resonance information are acquired simultaneously, in parallel, with one another, and the elements of those sets preferably are subsequently  
30 processed in parallel with one another to reconstruct images.

These, and other, features and advantages of the present invention constitute a generalization of, and an improvement upon, the SMASH apparatus and method. More specifically, the present invention contemplates the placement of an array of RF

coils, comprising substantially any number of RF receiver coils, at least partially around an object of interest located in the imaging volume of a magnetic imaging device. The present invention also contemplates the provision of a so-called "parallel" imaging capability wherein the output image may be taken in any plane transverse to the imaging volume of the apparatus.

In the present invention, parallel encoding in MRI is achieved by using the sensitivity profiles of an array of RF receiver coils at least partially surrounding the object of interest. Given this fact, the equation describing the MR signal seen by the  $i^{\text{th}}$  coil of a coil array surrounding the imaging volume of a magnetic resonance imaging device can be written as:

$$S_i(t) = \int \int \rho(x, y) \cdot W_i(x, y) \cdot e^{j\gamma(G_x x t + G_y y \tau)} dx dy$$

Where  $W_i(x, y)$  represents the 2D sensitivity profile of the  $i^{\text{th}}$  coil of the array, and  $\rho(x, y)$  is an image slice in a selected  $(x, y)$  plane. Then, taking the Fourier transform of that signal with respect to  $x$  yields:

$$F_i(x) = FT[S_i(t)] = \int \rho(x, y) \cdot W_i(x, y) \cdot e^{j\gamma G_y \tau} dy$$

The latter equation represents a projection of the phase modulated image  $\rho(x, y)$  onto the  $x$ -axis. Further, this signal can be represented in discrete form by the following matrix product:

$$F_i(x) = \sum_y [W_i(x, y) \cdot e^{j\gamma G_y \tau}] \cdot [\rho(x, y)]$$

If the number of receiver coils used simultaneously is  $N$ , and the 2D sensitivity profile  $W_i(x, y)$  of each one of them is

known, it is possible to reconstruct the image from only one  $N^{\text{th}}$  of the total number of k-space lines that would normally be required. Accordingly, the apparatus and method of the present invention use phase modulated projections of the received magnetic resonance data onto the frequency encoded (x-) axis, weighted by the 2D sensitivity profiles of the coils in the array, in order to reconstruct  $\rho(x,y)$  column by column (i.e., orthogonal to the x-axis).

The 2D sensitivity profiles are calculated first by acquiring a baseline image using the RF body coil of the conventional magnetic resonance imaging apparatus, whereby the sensitivity profile may be considered to be constant ( $W_b(x,y) = 1$ ) and a slice of the image in the (x, y) plane may be written as:  $\rho(x,y)$ . An image is then acquired with each coil in the array. In the  $i^{\text{th}}$  coil, this image can be written as:

$$W_i(x,y) \rho(x,y)$$

$W_i(x,y)$  is then computed by forming a point-by-point ratio between magnetic resonance data from the  $i^{\text{th}}$  coil and that from the body coil.

It, therefore, will be seen that a preferred embodiment of the invention is a method for generating a magnetic resonance image of an object of interest composed of a plurality of adjacent image lines. The method generally includes the following steps. First, a magnetic resonance imaging device is provided. This device typically includes a magnet system providing a background magnetic field in an imaging volume, central processor means, memory means, a RF coil surrounding the imaging volume, and a plurality of radio frequency receiver coils defining a multi-dimensional array thereof disposed about the imaging volume. A two-dimensional sensitivity profile for each receiver coil in the multi-dimensional array is then computed and recorded to the memory means. Thereafter, a plurality of magnetic resonance signals

of the object of interest located within the imaging volume is acquired from each receiver coil and recorded to the memory means. An image of the object of interest in a desired plane extending transversely through the imaging volume of the magnetic resonance imaging device then is reconstructed line-by line by the central processor means. This reconstruction combines the inverse of the matrix of the sensitivity profiles of each receiver coil and the matrix of the recorded MR data signals together to provide an image that may be displayed or printed.

Preferably, the method of the invention involves a calibration step wherein a homogeneous water phantom is provided to obtain the sensitivity profile of each receiver in the array. The homogeneous water phantom is located in the image volume of the magnetic resonance imaging device. Magnetic resonance data from the homogeneous water phantom in each plane of interest extending transversely through the imaging volume of the magnetic resonance imaging device then are acquired by a RF body coil and by each of the RF image receiver coils. This magnetic resonance data is stored in the memory of the magnetic resonance imaging device. Thereafter, the central processor means calculates, and stores in the memory means, the respective point-by-point ratios of the complex data representing the image from the water phantom provided by said receiver coils to the complex data representing the image from the water phantom provided by the RF body coil.

In a preferred embodiment of the invention, an object of interest is located in the imaging volume of the magnetic resonance imaging device. Magnetic resonance data from the plane of interest is acquired from each of the RF receiver coils and stored in memory. Then, the respective point-by-point ratios of the magnetic resonance data from the plane of interest provided by the receiver coils to the magnetic

resonance data from the water phantom are calculated so as to define a scaling factor for each point in the plane of interest. This scaling factor may then be utilized to provide a load-weighted, point-by-point sensitivity profile for each RF receiver coil in each plane of interest parallel to that for which the scaling factor was originally determined.

In still another embodiment of the invention, means are provided for determining and storing in memory the point-by-point time difference of signal reception by each of the receiver coils. This information may be combined with the sensitivity information in the inverse of the matrix of the sensitivity profile so as to provide a phase compensated, load-weighted, point-by-point, inverse, sensitivity profile matrix for each said receiver coil.

Further, it will be understood that the present invention includes apparatus for carrying out each of the above-described methods. Accordingly, the present invention also includes an apparatus for generating a magnetic resonance image of an object of interest - the image being composed of a plurality of adjacent image lines. More specifically, the apparatus preferably constitutes a magnetic resonance imaging device including an imaging volume, preprogrammed central processor means, memory means, a body coil surrounding said imaging volume, a plurality of radio frequency receiver coils defining a multi-dimensional array thereof disposed about said imaging volume, and display means. Each receiver coil in the multi-dimensional array has an associated two-dimensional sensitivity profile. Preferably, the preprogrammed central processing means is adapted to calculate each of the sensitivity profiles and to transfer the same to the memory means.

The receiver coils are adapted to acquire simultaneously a plurality of magnetic resonance signals from an object of interest located within the imaging volume, and to transfer



the same to the memory means. In addition, the central processor means is adapted to reconstruct, and to display on the display means, line-by-line, a two-dimensional image taken in a selected plane extending transversely through the object of interest.

This is accomplished by multiplying the inverse of the matrix of sensitivity profiles of the receiver coils and the matrix of data signals acquired by the receiver coils, and displaying the resultant product on the display means.

In another embodiment, the magnetic resonance imaging device is adapted to acquire, and to transfer to the memory means, magnetic resonance data from a homogeneous phantom in each plane of interest extending through the imaging volume using said body coil. Similarly, the magnetic resonance imaging device also is adapted to acquire, and to transfer to the memory means, magnetic resonance data from a homogeneous water phantom in each plane of interest from each of said receiver coils. Also, the central processing means is adapted to calculate, and transfer to the memory means, the respective point-by-point ratios of the magnetic resonance data from the water phantom provided by the receiver coils to the magnetic resonance data from the water phantom provided by the body coil. The component points of each image line may then be refined in the central processing means by a corresponding one of these ratios prior to its display on the display means.

In yet another embodiment, the magnetic imaging device is adapted to acquire, and to transfer to the memory means, magnetic resonance data from the object of interest by each of the receiver coils. In this case, the central processing means is adapted to calculate, and to transfer to the memory means, the respective point-by-point ratios of the magnetic resonance data from the object of interest provided by the receiver coils to the magnetic resonance data from the water phantom provided by the receiver coils. This ratio defines

for each point a scaling factor. The central processing means also is adapted to refine the component points of each image line by use of a corresponding one of the scaling factors prior to its display.

5        Finally, in still another embodiment of the invention, phase detection means are provided for determining and storing in the memory means the point-by-point time difference of signal reception by each of the receiver coils. In this alternative, the central processing means is adapted to  
10 multiply the point-by-point time difference of signal reception by the point-by-point sensitivity profile for each said surface coil prior to the line-by-line image display on the display means.

#### 15 BRIEF DESCRIPTION OF THE DRAWINGS

These and other features, objects and advantages of the present invention, will be more completely understood by those skilled in the art from the following detailed description of the preferred embodiments of the invention read in conjunction  
20 with the appended drawings. In the appended drawings, like reference numeral are used to refer to like elements throughout, and:

Fig. 1 is an illustrative depiction of a magnetic resonance imaging device in accordance with the present  
25 invention;

Fig. 1A is a schematic illustration of the disposition of the RF receiver coil array of the present invention disposed at least partially surrounding the imaging volume of the magnetic imaging device shown in Fig. 1;

30        Fig. 2 is a flow chart illustratively depicting the steps of the present method and their interrelationship with one another;

Fig. 3 is a schematic representation of the MRI acquisition and image reconstruction scheme of the present invention;

Fig. 4 is a schematic illustration of an exemplary one-dimensional version of the acquisition and image reconstruction scheme of the present invention;

Figs. 4A-4C are schematic illustrations of the sensitivity profiles denoted by  $W_1(i, j)$ ,  $W_2(i, j)$  and  $W_3(i, j)$ , respectively;

Fig. 5 is a composite image depicting simulation images received with a coil having the same sensitivity profile placed at 16 different positions about a brain;

Fig. 6 is a composite image showing the results of a reconstruction simulation using different sets of k-space lines;

Figs. 7A-7B illustratively depict reconstructions of a (256x256) image of a GE quality phantom using the body coil and receiver coils with a 16 fold speed enhancement; and

Figs. 8A-8B illustratively depict reconstructions of a brain image using the same techniques as the corresponding depictions of Figs. 7a and 7b, respectively.

#### DETAILED DESCRIPTION OF PREFERRED EMBODIMENT

As generally noted above, the present invention is implemented in a MRI setting. A parallel acquisition apparatus and method using 2D sensitivity maps of receiver coils disposed in an array at various arbitrary locations about an imaging volume in a magnetic resonance imaging apparatus is provided in order to encode the image space.

However, the invention allows for a more general positioning of the RF receiver coils, and can be used to create images derived from any plane in the imaging volume.

Further, the present invention has demonstrated an order of magnitude gain in imaging speed that varies, depending upon the number of coils used in the receiver array. In addition, an easy way of computing the 2D sensitivity profile of a RF receiver coil is provided using a sensitivity map model that may be acquired from a readily available homogeneous water phantom.

Hence, the apparatus and method of this invention are very flexible in that they can be used to speed up any pulse sequence. Indeed, it is anticipated that this novel apparatus and method will be found to be particularly well suited for cardiac imaging applications where fast spin-echo sequences are needed.

Referring now to the drawings, and particularly to Figs. 1 and 1A, it will be seen that the present invention provides a magnetic resonance imaging system that is basically of the type well known in the art. The system includes a plurality of gradient coils that produce spatially encoded gradients imposed upon a background magnetic field  $B_0$  within the volume in which the object to be examined is placed. In addition, an array of RF receiver coils is disposed in circumferentially spaced relation to one another about the imaging volume.

More specifically, the magnetic resonance imaging system, generally indicated at 2, may include a Helmholtz coil pair 4 used to generate a large, static, substantially homogeneous magnetic field in the imaging space or volume 6 in a direction parallel to the z-axis or the line 8. An object or subject (not shown) may be placed in the imaging volume within the cylinder 10 for examination using NMR phenomena. The subject is placed on the z-axis or the line 8, and is located within the coil 12. The coil 12 is representative of devices used to generate RF fields in the subject placed in the system for examination.

These RF fields, when in the presence of a static magnetic field  $B_0$ , cause the occurrence of magnetic resonance in the nuclei of certain elements, such as hydrogen, in the specimen to be examined.

5       The direction of the static magnetic field ( $B_0$ ) produced by the coil pair 4 is indicated in Fig. 1 by an arrow. Currents are made to flow in the RF coil 12. The direction of the currents reverse each half-cycle of the alternating RF current in coil 12. This produces a transverse magnetic field  
10 of low magnitude compared to  $B_0$ . The magnitude of the flux density resulting from the static magnetic-field intensity  $B_0$  may be typically on the order of Tesla.

The static magnetic field  $B_0$  is constant while the subject is in the system for analysis or examination. The RF  
15 transverse magnetic field is applied for a time sufficient only to allow the protons in the hydrogen atoms (or the nuclei of other atoms exhibiting the magnetic resonance phenomenon) to be affected such that precession of the net magnetization of the subject occurs. The precession of the net magnetic  
20 field associated with the nuclei in the subject occurs at the Larmor frequency, which is directly proportional to the magnitude of the magnetic field at the location of the nuclei. This can be detected as an NMR signal which provides the information for reconstruction of an image.

25       In MRI systems, it is necessary to provide various gradient coils (not shown) for producing spatially encoding gradients that are imposed upon the static magnetic field within the region in which the subject to be examined is placed also are provided. The gradient-coil apparatus is  
30 typically positioned on the outside of a cylindrical surface, such as the surface 10, which may be used as a support structure for the gradient coils. The gradient coils conventionally produce linear magnetic field gradients in any of the three orthogonal directions  $x$ ,  $y$  and  $z$ .

Conventionally, the direction of the line 8 is designated as the z-direction or z-axis and the x- and y-axes of the coordinate system are orthogonal to the z-axis and to one another.

5       The typical configuration of a z-gradient coil is illustrated in U.S. Pat. No. 4,468,622, the disclosure of which is hereby incorporated by reference. The configuration of the typical transverse (x or y) gradient coils is illustrated in U.S. Pat. No. 4,486,711, the disclosure of  
10       which also is hereby incorporated by reference.

      The present invention also contemplates that an array of RF receiver coils 14 will surround (or partially surround) the imaging volume (see Fig. 1A). These RF receiver coils 14 provide the required system calibration and input information  
15       necessary to enhance the speed of parallel magnetic resonance data acquisition and parallel image reconstruction, without serious adverse effect upon acceptable signal-to-noise ratios. More simply, the concept of parallel imaging in the present invention is based on using multiple receiver coils, with each  
20       coil providing independent information about the image.

      Fig. 2 is a flow diagram illustratively showing the basic sequence of the method of the invention. Thus, it will be seen that the present method contemplates that an apparatus such as that described above will be provided. In this  
25       apparatus, the CPU is preprogrammed to carry out numerous operations upon signals received in parallel from various coils of the receiver array, and to output an image to the display means. Among these numerous operations are the following: (1) the computation of the 2D sensitivity profile  
30       of each coil of the receiver array; (2) storing the inverse of the calculated sensitivity profile matrix of each coil of the receiver array in memory; and (3) reconstructing an image taken in a selected plane extending transversely through the imaging volume by calculating and transmitting to the display

means in parallel with one another the various display lines that make up the entire image.

The theory behind this method has been alluded to above. Specifically, an MR signal received in a receiving coil having  
 5  $W_k(x, y)$  as its two-dimensional sensitivity profile, when neglecting T2 decay, can be written as:

$$s_k(G_y^g, t) = \int \int \rho(x, y) W_k(x, y) e^{i\gamma(G_x x t + G_y^g y \tau)} dx dy$$

10

Where  $\rho(x, y)$  denotes the proton density function,  $G_x$  represents the readout gradient amplitude applied in the  $x$  direction,  $G_y^g$  represents the phase encoding gradient applied during the  $g^{\text{th}}$  acquisition,  $x$  and  $y$  represent the  $x$  and  $y$   
 15 positions respectively, and  $\tau$  is the pulse width of the phase decoding gradient  $G_x$ .

In most conventional serial imaging sequences, the phase decoding gradient is constant along one direction, and the phase encoding is applied along an orthogonal direction. In  
 20 addition, only one receiver coil is used to collect all the data required to reconstruct  $\rho(x, y)$ . To achieve that, the phase encoding gradient  $G_y$  is varied so as to cover all of  $k$ -space with the desired resolution. For each value of  $G_y^g$  of  $G_y$ , an echo is acquired. This makes serial imaging a time-  
 25 consuming procedure. In this technique, however, sensitivity profile information from a number of receiver coils is used in order to minimize the number of acquisitions needed to estimate and reconstruct  $\rho(x, y)$ . Taking the Fourier Transform of equation (1) along the  $x$  direction when a phase  
 30 encoding gradient  $G_y^g$  is applied yields:

$$S_k(G_y^g, x) = \int \rho(x, y) W_k(x, y) e^{j\gamma(G_y^g y \tau)} dy,$$

This is the phase-modulated projection of the sensitivity-weighted image onto the x-axis. A two-dimensional orthogonal spatial basis function denoted by  $\Omega(x, y)$  is used to expand the quantities in equation (2) for the purpose of discretizing it. A constant spatial sampling dwell of  $\Delta_s$  is assumed, and the following change of variables is made:

$$x = m\Delta_s \text{ and } y = n\Delta_s.$$

Then, performing the expansion in terms of  $\Omega(x, y)$  yields:

$$S_k(G_y^g, m\Delta_s) = \int \Sigma_n \rho(m\Delta_s, n\Delta_s) \cdot W_k(m\Delta_s, n\Delta_s) \cdot e^{i\tau(G_y^g n\Delta_s)} \Omega(x - m, y - n) dy \quad (4)$$

Removing  $\Delta_s$  and rearranging the terms for simplification, results in:

$$S_k(G_y^f, m) = \Sigma_n \rho(m, n) \cdot W_k(m, n) \cdot e^{i\tau(G_y^f n)}$$

This expression can be converted to matrix form for each position  $m$  along the horizontal direction of the image, as follows:

$$\begin{pmatrix} S_1(G_y^1, m) \\ S_1(G_y^f, m) \\ S_2(G_y^1, m) \\ S_2(G_y^f, m) \\ \vdots \\ S_k(G_y^1, m) \\ S_k(G_y^f, m) \end{pmatrix} = \begin{pmatrix} W_1(m, 1) \cdot \phi(G_y^1, 1) & \dots & W_1(m, n) \cdot \phi(G_y^1, n) \\ W_1(m, 1) \cdot \phi(G_y^f, 1) & \dots & W_1(m, n) \cdot \phi(G_y^f, n) \\ W_2(m, 1) \cdot \phi(G_y^1, 1) & \dots & W_2(m, n) \cdot \phi(G_y^1, n) \\ W_2(m, 1) \cdot \phi(G_y^f, 1) & \dots & W_2(m, n) \cdot \phi(G_y^f, n) \\ \vdots & \vdots & \vdots \\ W_k(m, 1) \cdot \phi(G_y^1, 1) & \dots & W_k(m, n) \cdot \phi(G_y^1, n) \\ W_k(m, 1) \cdot \phi(G_y^f, 1) & \dots & W_k(m, n) \cdot \phi(G_y^f, n) \end{pmatrix} \cdot \begin{pmatrix} \rho(m, 1) \\ \rho(m, 2) \\ \rho(m, 3) \\ \vdots \\ \vdots \\ \rho(m, n) \end{pmatrix}$$

Where  $f$  is the number of phase encodes used in the experiment,



$\phi(G_y^g, p) = e^{i\gamma(G_y^g p \tau)}$  for every position  $p$  along the  $y$  axis and every phase encoding gradient value  $G_y^g$ . Equation (6) can be written in a more compact form as follows:

$$[S_k(G_y^g, m)] = [W_k(m, n) \cdot e^{i\gamma(G_y^g(n)\tau)}] \cdot [\rho(m, n)]$$

Where  $[S_k(G_y^g, m)]$  represents a matrix of dimensions  $(K \times N)$ , and  $K$  is the product of the number of coils  $k$  in the array by the number of phase encodes acquired  $f$ , which is equal to the number of echoes collected in all the coils combined;  $M$  is the number of time points per echo, which is equal to the number of pixels resolved in the  $x$  direction, and  $N$  is the desired number of pixels in the  $y$  direction. For square image matrices,  $M$  must equal  $N$ .

The matrix  $[A_m(k, y)] = [W_k(m, n) \cdot e^{i\gamma(G_y(n)\tau)}]$  is three-dimensional with dimensions  $(K \times M \times N)$ , and  $[p(m, n)]$  represents the image of size  $(M \times N)$ . If the matrix index  $m$  is fixed to a value  $m_j$ , the size of  $[A_{m_j}]$  then becomes  $(K \times N)$ . Hence, in order to reconstruct the image  $[p(m, n)]$ , a matrix pseudoinverse of  $[A_{m_j}(k, y)]$  is computed for every position  $m_j$  along the  $x$ -axis. This yields a column by column reconstruction. The choice of  $K$  affects the quality of the reconstruction. Practically, increasing  $K$  would result in an increase of the rank of the matrix  $[A_{m_j}(k, y)]$ , yielding a pseudoinverse that is better conditioned. Equation 10 is schematically represented in Fig. 3 for clarity.

In addition, a simplified version of the above concepts is depicted in Fig. 4, and described hereinbelow, to further clarify the present method.

Assume the image to be of size  $3 \times 3$  as shown in Fig. 4, where each pixel has an intensity value denoted by  $r(i, j)$ , where  $i$  is the column number and  $j$  is the row number. The goal is to determine and reconstruct the values of  $r(i, j)$  for all the pixels of the image. Further assume that three coils

1, 2 and 3 are arrayed around the image, and are used for image acquisition as shown in Fig. 4. These coils have sensitivity profiles denoted by  $W_1(i,j)$ ,  $W_2(i,j)$ ,  $W_3(i,j)$ , respectively. (see Figs.4A-4C)

5 The sensitivity profiles of the coils are determined from a homogeneous image by placing a water phantom whose pixel intensity values are equal to  $I(i,j)$  into the imaging volume, and acquiring its image using any chosen pulse sequence with the body coil whose sensitivity profile  $W_b(i,j) = 1$  for all the  
 10 pixels in the image. The reconstructed image would have pixel intensity values of  $r_b(i,j) = I(i,j)$  for all pixels. Next, images are acquired using the 3 coils shown in Fig. 4 with respective sensitivity profiles  $[W_1(i,j)]$ ,  $[W_2(i,j)]$ ,  $[W_3(i,j)]$  to get three images of pixel intensity values equal to  
 15  $[I(i,j) \cdot W_1(i,j)]$ ,  $[I(i,j) \cdot W_2(i,j)]$ ,  $[I(i,j) \cdot W_3(i,j)]$ , respectively. Taking the point-by-point ratio of the images acquired using the three coils in the array divided by the RF body coil image yields the sensitivity profiles  $W_1(i,j)$ ,  $W_2(i,j)$ , and  $W_3(i,j)$ , respectively.

20 Thereafter, the homogeneous water phantom is replaced with the object of interest to be imaged. This object has pixel intensities  $r(i,j)$  as shown in Figure 4. The MR signal received in coil k can be written as:

25 
$$S_k(t) = \int \int r(x,y) \cdot W_k(x,y) \cdot e^{j\gamma(G_x x + G_y y)} dx \cdot dy$$

Where  $G_x$  and  $G_y$  represent the gradients applied in the x and y directions respectively. Then taking the fourier transform  
 30 with respect to x provides:

$$s_k(x) = \int r(x,y) \cdot W_k(x,y) \cdot e^{j\gamma(G_y y)} dy$$

This is the phase modulated projection of the image  $r(x,y)$  weighted by the sensitivity profile  $W_k(i,j)$  onto the x-axis. Discretizing, wherein the index  $i$  corresponds to a discrete x position and the index  $j$  corresponds to a discrete y position, provides:

$$s_k(i) = \sum_j r(i,j) \cdot W_k(i,j) \cdot e^{j\gamma(G_y j \tau)}$$

whereby in coil 1,

$$s_1(i) = r(i,1) \cdot W_1(i,1) \cdot e^{j\gamma(G_y 1 \tau)} + r(i,2) \cdot W_1(i,2) \cdot e^{j\gamma(G_y 2 \tau)} + r(i,3) \cdot W_1(i,3) \cdot e^{j\gamma(G_y 3 \tau)}$$

In coil 2,

$$s_2(i) = r(i,1) \cdot W_2(i,1) \cdot e^{j\gamma(G_y 1 \tau)} + r(i,2) \cdot W_2(i,2) \cdot e^{j\gamma(G_y 2 \tau)} + r(i,3) \cdot W_2(i,3) \cdot e^{j\gamma(G_y 3 \tau)}$$

and in coil 3,

$$s_3(i) = r(i,1) \cdot W_3(i,1) \cdot e^{j\gamma(G_y 1 \tau)} + r(i,2) \cdot W_3(i,2) \cdot e^{j\gamma(G_y 2 \tau)} + r(i,3) \cdot W_3(i,3) \cdot e^{j\gamma(G_y 3 \tau)}$$

These equations represent the modulated projections of the image onto the x axis. They also are three linear equations with three unknowns

$[r(i,1), r(i,2), r(i,3)]$  for every position  $i$  along the x axis.

Recalling that the goal is to solve for the pixel intensity values of the vector  $[r(i,1), r(i,2), r(i,3)]$  which is a column of the image at position  $i$  along the x axis, it is noted that the following equation may be written for each position  $i$  along the x axis:

$$\begin{pmatrix} s_1(i) \\ s_2(i) \\ s_3(i) \end{pmatrix} = \begin{pmatrix} W_1(i,1) \cdot e^{j\gamma(G_y 1 \tau)} & W_1(i,2) \cdot e^{j\gamma(G_y 2 \tau)} & W_1(i,3) \cdot e^{j\gamma(G_y 3 \tau)} \\ W_2(i,1) \cdot e^{j\gamma(G_y 1 \tau)} & W_2(i,2) \cdot e^{j\gamma(G_y 2 \tau)} & W_2(i,3) \cdot e^{j\gamma(G_y 3 \tau)} \\ W_3(i,1) \cdot e^{j\gamma(G_y 1 \tau)} & W_3(i,2) \cdot e^{j\gamma(G_y 2 \tau)} & W_3(i,3) \cdot e^{j\gamma(G_y 3 \tau)} \end{pmatrix} \cdot \begin{pmatrix} r(i,1) \\ r(i,2) \\ r(i,3) \end{pmatrix}$$

Subsequently, solving for the vector

$$\begin{pmatrix} r(i, 1) \\ r(i, 2) \\ r(i, 3) \end{pmatrix}$$

5 one takes the inverse of the matrix

$$\begin{pmatrix} W_1(i, 1).e^{j\gamma(G_v 1\tau)} & W_1(i, 2).e^{j\gamma(G_v 2\tau)} & W_1(i, 3).e^{j\gamma(G_v 3\tau)} \\ W_2(i, 1).e^{j\gamma(G_v 1\tau)} & W_2(i, 2).e^{j\gamma(G_v 2\tau)} & W_2(i, 3).e^{j\gamma(G_v 3\tau)} \\ W_3(i, 1).e^{j\gamma(G_v 1\tau)} & W_3(i, 2).e^{j\gamma(G_v 2\tau)} & W_3(i, 3).e^{j\gamma(G_v 3\tau)} \end{pmatrix}$$

and multiplies it with the vector

10

$$\begin{pmatrix} s_1(i) \\ s_2(i) \\ s_3(i) \end{pmatrix}$$

with the result that:

15

$$\begin{pmatrix} r(i, 1) \\ r(i, 2) \\ r(i, 3) \end{pmatrix} = \begin{pmatrix} W_1(i, 1).e^{j\gamma(G_v 1\tau)} & W_1(i, 2).e^{j\gamma(G_v 2\tau)} & W_1(i, 3).e^{j\gamma(G_v 3\tau)} \\ W_2(i, 1).e^{j\gamma(G_v 1\tau)} & W_2(i, 2).e^{j\gamma(G_v 2\tau)} & W_2(i, 3).e^{j\gamma(G_v 3\tau)} \\ W_3(i, 1).e^{j\gamma(G_v 1\tau)} & W_3(i, 2).e^{j\gamma(G_v 2\tau)} & W_3(i, 3).e^{j\gamma(G_v 3\tau)} \end{pmatrix}^{-1} \begin{pmatrix} s_1(i) \\ s_2(i) \\ s_3(i) \end{pmatrix}$$

20

Hence, the image may be reconstructed column by column in parallel.

As described above, the inventive method is based on using the sensitivity profiles of RF receiver coils to encode MR data. As suggested in the above simplified example, it is possible to calculate these profiles by first acquiring a baseline image of a homogeneous water phantom using a RF coil with a homogeneous sensitivity profile covering the whole image. This image can be written as  $I(x, y)$ . Subsequently, images of the same water phantom are acquired using the receiver coils. The image acquired using the  $k^{th}$  receiver coil can be represented as  $I(x, y).W_k(x, y)$  where  $W_k(x, y)$  is the sensitivity profile of the surface coil.

Then, taking the point-by-point ratio of the two images yields the sensitivity profile  $W_k(x,y)$ . It, however, must be understood that the sensitivity profiles of the receiver coils also depends upon their loading.

5 In the present invention, the variations incurred by these profiles amounts to a constant scaling between any two different loads. Consequently, to find the actual sensitivity profile of the receiver coils when loaded with an arbitrary object of interest, sensitivity profile estimation on that  
10 object as described above is performed. Then, a comparison (ratio) of those sensitivity profiles to the sensitivity profiles computed on the basis of a homogeneous water phantom is made in order to extract the scaling factor. The sensitivity profiles calculated from the homogeneous water  
15 phantom are then multiplied by the scaling factor so computed, and the result is used in the encoding scheme. This is done in order to get the maximum coverage possible of the field of view by the sensitivity profile. Other methods have been described in the literature for performing a sensitivity  
20 profile calculation. However, the technique described and illustrated above appears to be particularly well suited for our reconstruction method. This is particularly true since the method is not time-consuming, and it can be carried out essentially in real time.

25 In order to appropriately account for all of the frequencies included in the image, the choice of the phase modulations used in the inversion matrix should be determined by the frequency content of the sensitivity profile. In most cases, the sensitivity profiles of RF coils have low frequency  
30 content. Accordingly, high frequency phase modulations should be used to well represent the image. This means that appropriate k-space lines can be chosen by those skilled in the art for an optimal reconstruction.

#### Examples

Experiments were performed on a 1.5T GE SIGNA clinical MRI system at the 5.4 hardware-software configuration. The purpose of these experiments was to test the performance of the present methodology. In the experiments, image resolution was 256x256 pixels. A 16-echo RARE sequence, TR=2sec, TE=30ms per shot, were utilized to obtain one dynamic image update with a single shot.

In order to assess the technique herein described, simulations were performed.

10 In one case, a homogeneous MR data of a brain first was acquired using a head coil. Then, two further sets of MR data were acquired of a homogeneous water phantom - one using the head coil, and the other using a surface coil. The sensitivity profile of the receiver coil was then computed by taking the point-by-point ratio of the water phantom data  
15 acquired using the receiver coil over the data acquired using the head coil, as described above. Thereafter, sensitivity profiles were computed for RF receiver coils disposed about the object of interest with an angle of  $2\pi/16$  between any  
20 adjacent two of them. Next, the computed sensitivity profiles were multiplied point-by-point with the brain data to get approximations of sensitivity weighted images from surface coils placed at different positions. These images are shown in Fig.5.

25 The sensitivity-weighted images were then Fourier Transformed in two dimensions in order to get the relevant k-space data. Subsequently, 16 lines of k-space data were taken from each matrix, and those lines were used to reconstruct the head image using our technique. As mentioned, the choice of  
30 the k-space lines affects the image resolution. In Fig.6, four images acquired using different sets of k-space data are shown. The image on the left shows a reconstruction using the 16 lines of k-space going between  $k=-7$  and  $k=8$ ; the second image from the left shows a reconstruction using the 16 lines

of k-space going between  $k=-39$  and  $k=40$  and acquiring every 5<sup>th</sup> line . The third image from the left shows a reconstruction using the lines of k-space going between  $k=-79$  to  $k=80$ , and acquiring every 10<sup>th</sup> line.

5

Finally, the image on the right shows a reconstruction using the 16 lines of k-space going between  $k=-127$  to  $k=128$ , and acquiring every 16<sup>th</sup> line. It can be seen that the third image from the left represents the best result of the four shown.

10 Further, a certain deterioration of the image is observed as more lines are skipped before reconstruction. An optimal combination of k-space lines can be found to achieve the desired result.

In another experiment, baseline data of the human head  
15 was acquired using the head coil. Then, using a receiver coil, 16 data sets from the human head taken at different coil positions were acquired. These data sets then were used in conjunction with the baseline data to calculate the sensitivity profiles of the coil at different positions using  
20 the point-by-point ratio method described above. These sensitivity profiles were then used in order to reconstruct images using only 16 lines of k-space from each coil. The results of these experiments are shown respectively in Fig.7 and Fig.8.

25 Accordingly, it will be understood that a parallel imaging and reconstruction method has been designed and implemented which is intended to substantially increase imaging speed in MRI. This technique is based on using 2D sensitivity profile maps derived from an array of receiver  
30 coils surrounding an imaging volume in order to encode and reconstruct MRI data in parallel. Therefore, imaging time may be reduced in proportion to the number of receiving coils present in the array. Further, this technique is adaptable to

all imaging sequences, and to all imaging planes, thereby making it very practical in a clinical setting.

Still further, the placement of the RF receiving coils may be arbitrary, around the imaged field of view. This  
5 allows for the placement of more RF coils in the receiver array, which in turn enables faster imaging. Nevertheless, since there is a tradeoff between coil size and penetration, an optimal number of coils can be computed to yield the best results.

10 Still further, in most parallel imaging schemes, decoupling of the RF receiver coils in the array is a necessary step before imaging. In the present invention, however, the coupling information between the different coils is incorporated into the reconstruction algorithm. This makes  
15 it easier to design and execute arrays that contain a large number of coils.

In the foregoing discussion, the sensitivity profiles of the respective receiver coils have been suggested to be independent of the coil loading. This permits an  
20 extrapolation of the results of the sensitivity profiles computed from a water phantom to be used in the encoding of MR data acquired from other objects. In reality however, loading affects the sensitivity profiles of the coil elements. Accordingly, the dynamic method of sensitivity profile  
25 adjustment for each different object to be imaged required in the course of the generation of images having practical utility.

It is to be understood that this description of certain presently preferred embodiments of the present invention has  
30 been provided by way of illustration, and not limitation. Various alterations, changes, variations, modifications and the like encompassed within the scope of the present invention in its broadest aspects will occur to those skilled in the art in view of the teachings and descriptions contained in this



specification. For example, the array of receiver coils may be used to emit the RF signal to eliminate the need for a RF body coil. Accordingly, the scope, content and limitations of the present invention are to be determined solely from the  
5 appended claims.

Claims:

What is claimed is:

1. A method for generating a magnetic resonance image  
5 of an object of interest wherein said image consists of a plurality of adjacent image lines, said method comprising the steps of:
- a) providing a magnetic resonance imaging device comprising a magnet system providing a homogeneous magnetic  
10 background field in an imaging volume, a RF body coil surrounding said imaging volume, and a plurality of radio frequency receiver coils defining a multi-dimensional array thereof disposed about said imaging volume;
  - b) determining a two-dimensional sensitivity profile  
15 for each said receiver coil in said multi-dimensional array;
  - c) acquiring simultaneously a plurality of magnetic resonance signals from said object of interest located within said imaging volume from each said receiver coil;
  - d) reconstructing a two-dimensional image taken in a  
20 selected plane extending through said object of interest using said two-dimensional sensitivity profile and said plurality of magnetic resonance signals.
- 25 2. The method according to Claim 1, wherein:  
step (b) further comprises:
- (i) providing a homogeneous water phantom;
  - (ii) locating said homogeneous water phantom in said  
image volume;
  - 30 (iii) acquiring magnetic resonance data from said homogeneous water phantom in each plane of interest extending transversely through said imaging volume using said RF body coil;

- (iv) using a two-dimensional Fourier Transformation to provide complex image data representing an image of said homogeneous water phantom in each said plane for the body coil;
- 5 (v) acquiring magnetic resonance data from said homogeneous water phantom in each plane of interest extending transversely through said imaging volume using each of said RF receiver coils;
- 10 (vi) using a two-dimensional Fourier Transformation to provide complex image data representing an image of said homogeneous water phantom in each said plane for each receiver coil; and
- 15 (vii) determining the respective point-by-point ratios of said complex image data from said water phantom provided by said receiver coils to said magnetic resonance data from said water phantom provided by said body coil, thereby obtaining a sensitivity map for each receiver
- 20 coil.

3. The method according to Claim 2, wherein step (b) further comprises:

- 25 (viii) locating an object of interest in said imaging volume;
- (ix) acquiring magnetic resonance data from said object in each plane of interest extending transversely through said imaging volume using
- 30 said RF body coil;
- (x) using a two-dimensional Fourier Transformation to provide complex image data representing an image of said object in each said plane for the body coil;

(xi) acquiring magnetic resonance data from said object in each plane of interest extending transversely through said imaging volume using each of said RF receiver coils;

5 (xii) using a two-dimensional Fourier Transformation to provide complex image data representing an image of said object in each said plane for each receiver coil; and

10 (xiii) calculating in said central processor and storing in said memory the respective point-by-point ratios of said complex image data from said object of interest provided by said receiver coils to said complex image data of said object of interest from said body coil to  
15 define for each said point a scaling factor.

4. An apparatus for generating a magnetic resonance image of an object of interest, wherein said image is composed of a plurality of adjacent image lines; said apparatus comprising:

a magnetic resonance imaging device comprising a magnet system providing a homogeneous magnetic background field in an imaging volume, preprogrammed central processor means, memory means, a RF body coil surrounding said imaging volume, a plurality of radio frequency receiver coils defining a multi-dimensional array thereof disposed about said imaging volume, and display means;

wherein each said receiver coil in said multi-dimensional array has an associated two dimensional sensitivity profile; said preprogrammed central processing means is adapted to calculate each said sensitivity profile and to transfer same to said memory means;

wherein said receiver coils are adapted to acquire simultaneously a plurality of magnetic resonance signals of said object of interest located within said imaging volume and to transfer same to said memory means; and

wherein said central processor means is adapted to reconstruct, and to display on said display means, line-by-line, a two dimensional image taken in a selected plane extending transversely through said object of interest by using said sensitivity profile of each said coil and corresponding signals from each of said receiver coils.

5. An apparatus according to Claim 5, wherein said magnetic resonance imaging device further is adapted to acquire and to transfer to said memory means:

(i) magnetic resonance data from a homogeneous water phantom in each plane of interest extending transversely through said imaging volume using said body coil, and

(ii) magnetic resonance data from said homogeneous water phantom in each said plane of interest from each of said receiver coils;

said central processing means is adapted

(1) to calculate and transfer to said memory means the respective point-by-point first ratios of complex image data of said water phantom provided by said receiver coils to corresponding complex image data of said water phantom provided by said body coil.

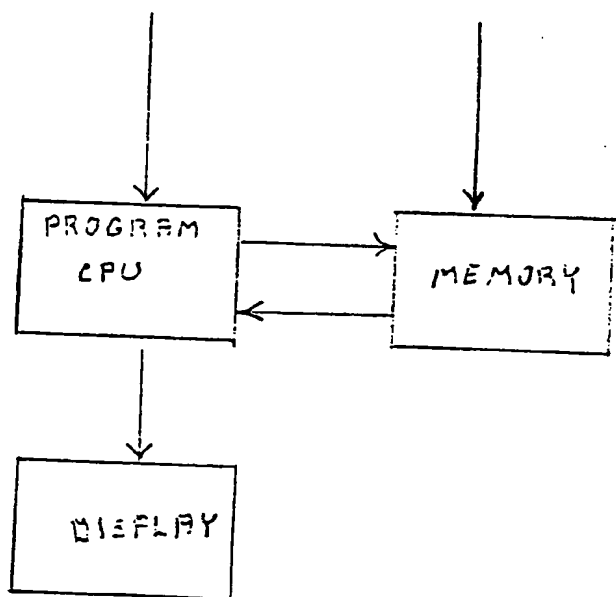
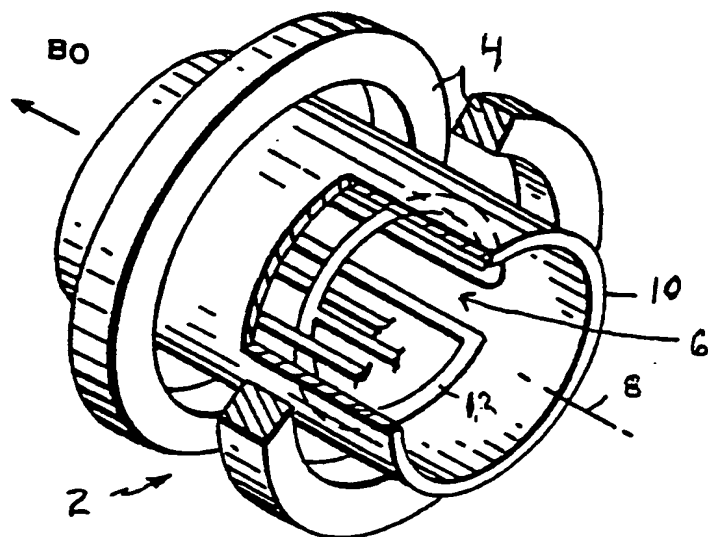


FIG.1

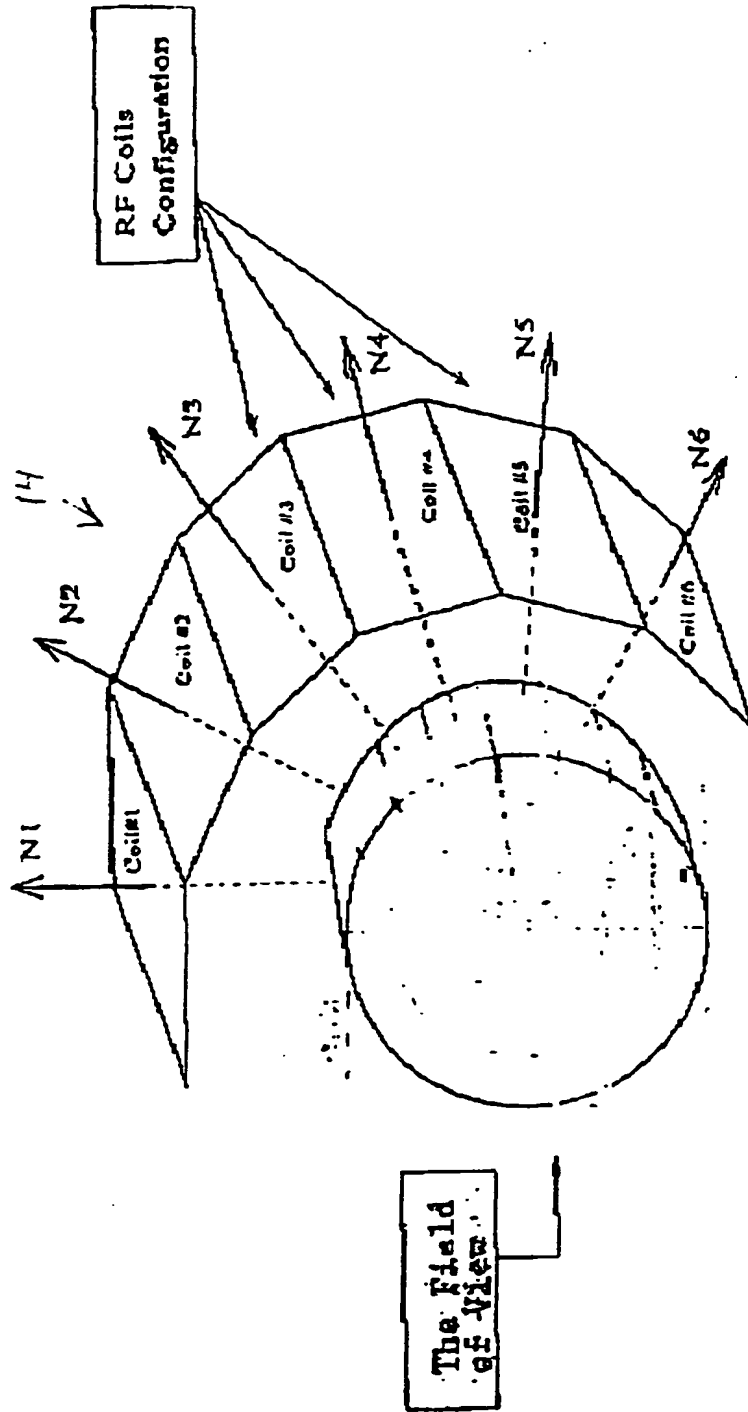


Figure 1A Schematic representation in 3-D of the Field of View and the RF pickup coil array positioning in this method. The coil orientations are described by vectors orthogonal to the coils.



3/10

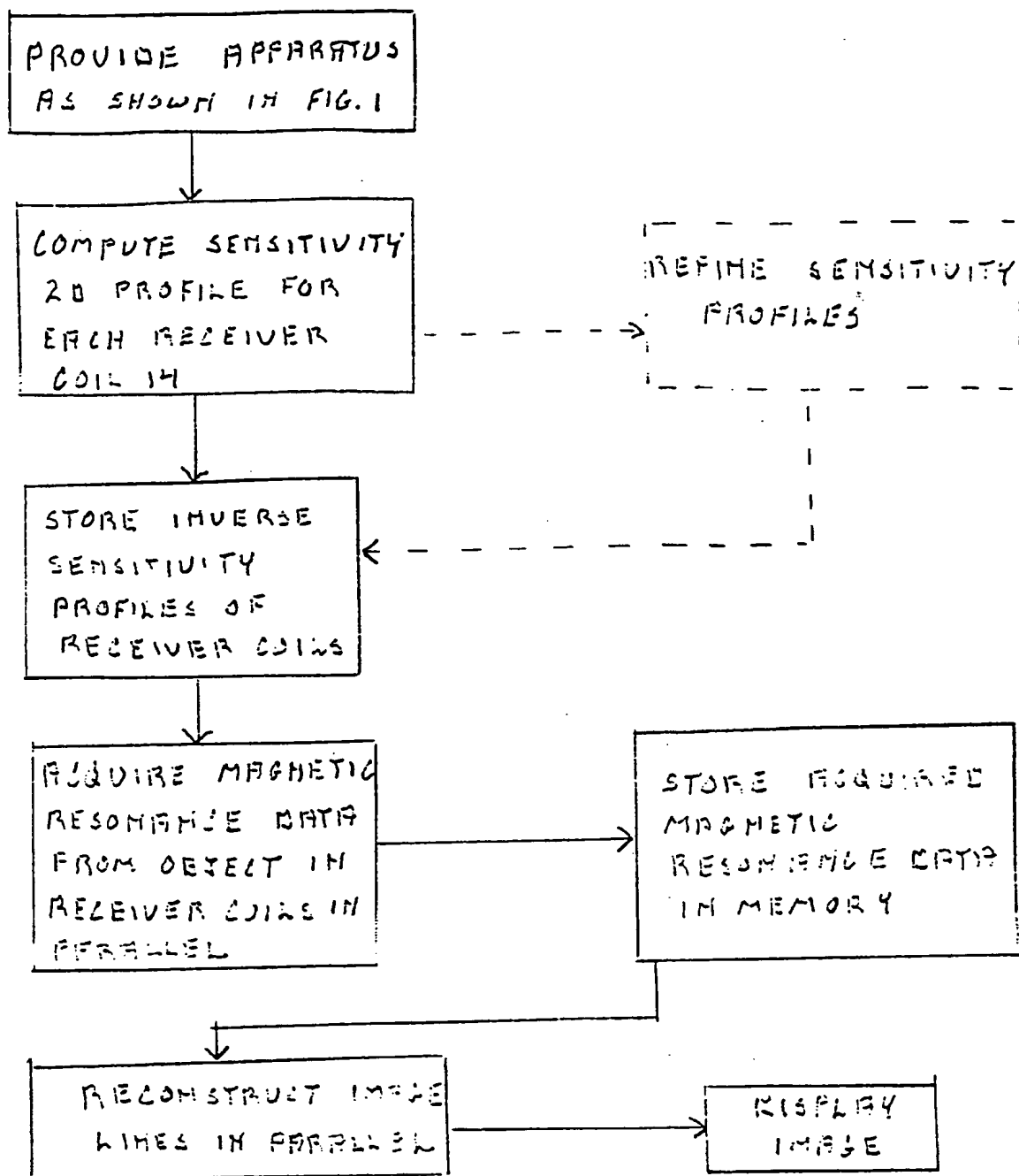


Fig. 2

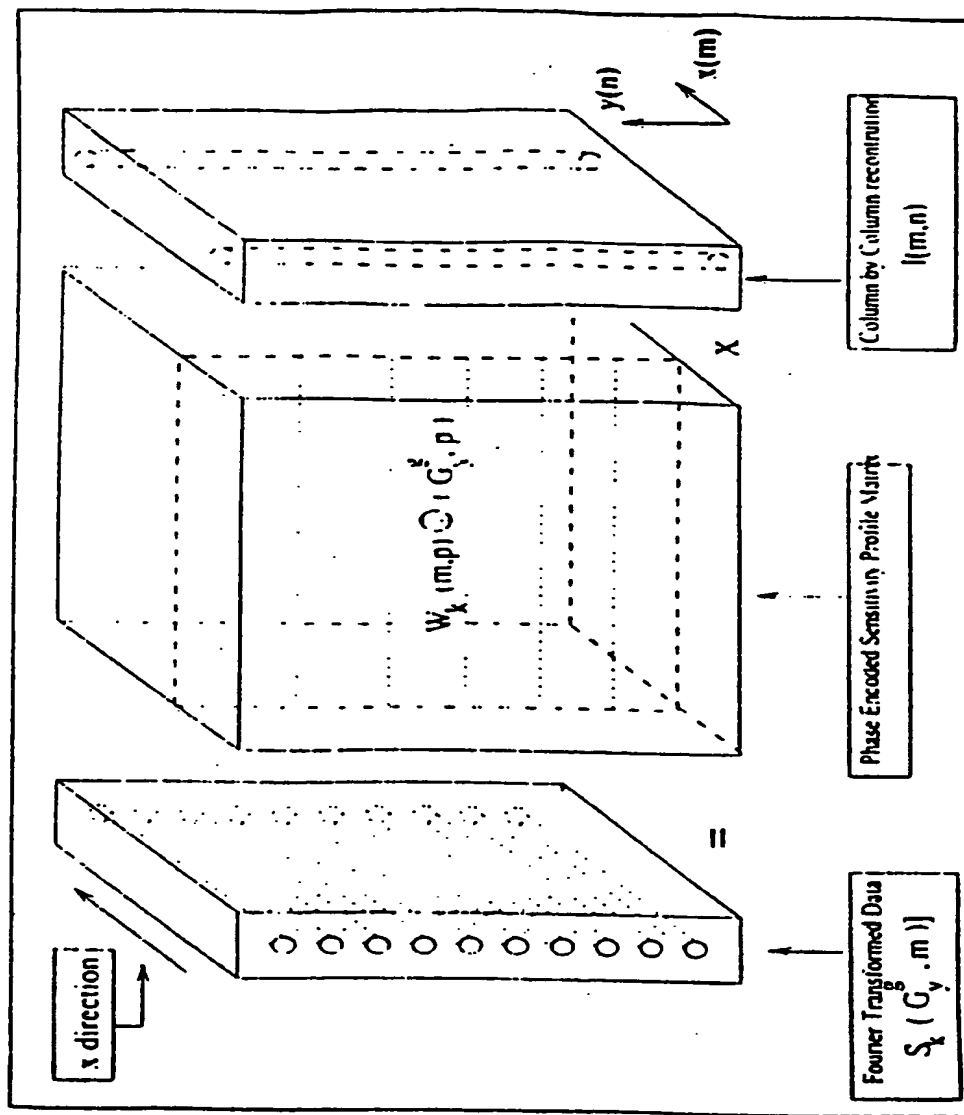


Figure 3 Schematic representation of the parallel reconstruction scheme. The matrix to the left represents the 1DFT of the k-space data chosen for reconstruction, the matrix in the center is the three dimensional sensitivity matrix referred to by A in the text, and the matrix on the right represents the image to be reconstructed.

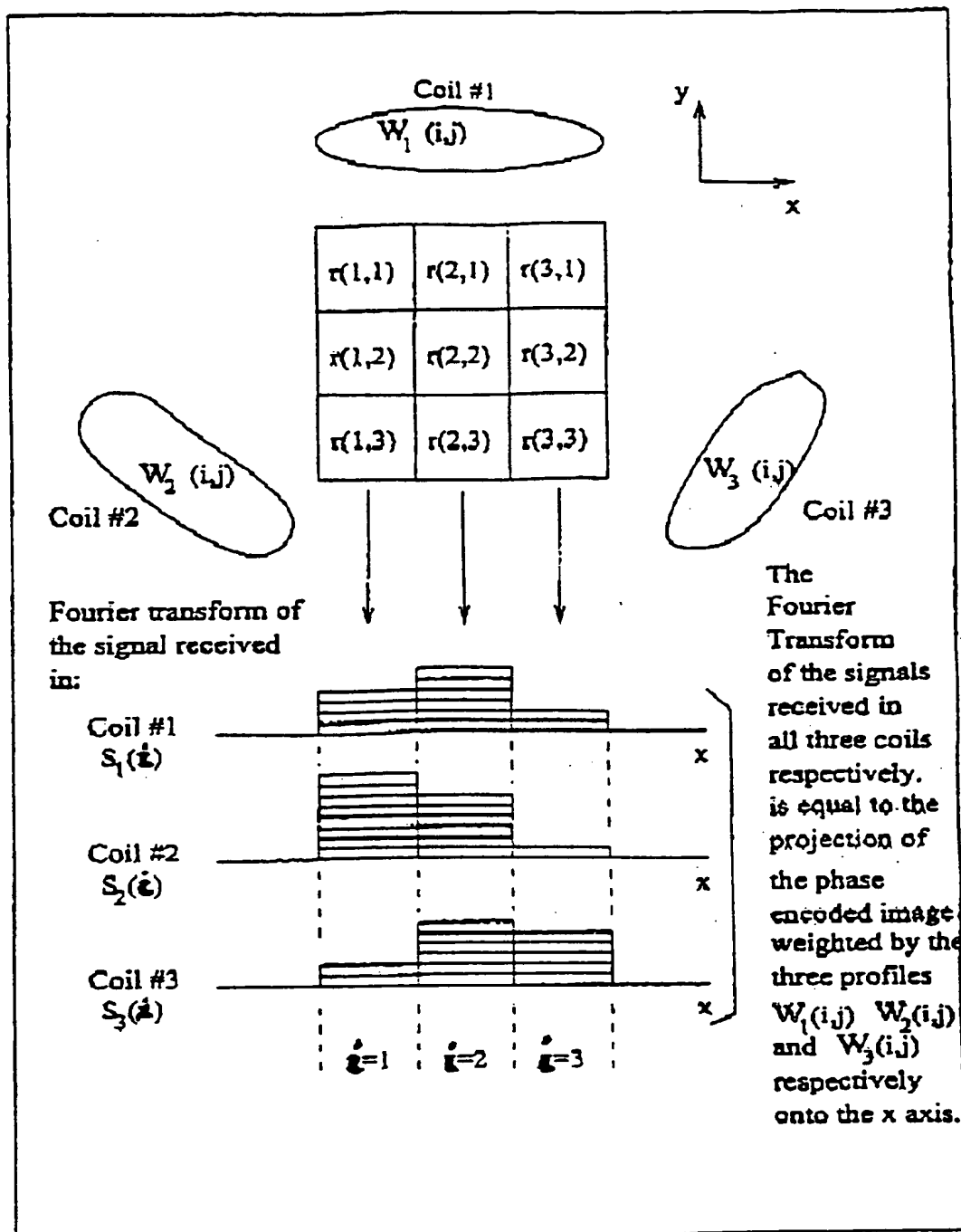


Figure 4

Coil #1

$W_1(1,1)$	$W_1(1,2)$	$W_1(1,3)$
$W_1(2,1)$	$W_1(2,2)$	$W_1(2,3)$
$W_1(3,1)$	$W_1(3,2)$	$W_1(3,3)$

FIG. 4A

Coil #2

$W_2(1,1)$	$W_2(1,2)$	$W_2(1,3)$
$W_2(2,1)$	$W_2(2,2)$	$W_2(2,3)$
$W_2(3,1)$	$W_2(3,2)$	$W_2(3,3)$

FIG. 4B

Coil #3

$W_3(1,1)$	$W_3(1,2)$	$W_3(1,3)$
$W_3(2,1)$	$W_3(2,2)$	$W_3(2,3)$
$W_3(3,1)$	$W_3(3,2)$	$W_3(3,3)$

FIG. 4C

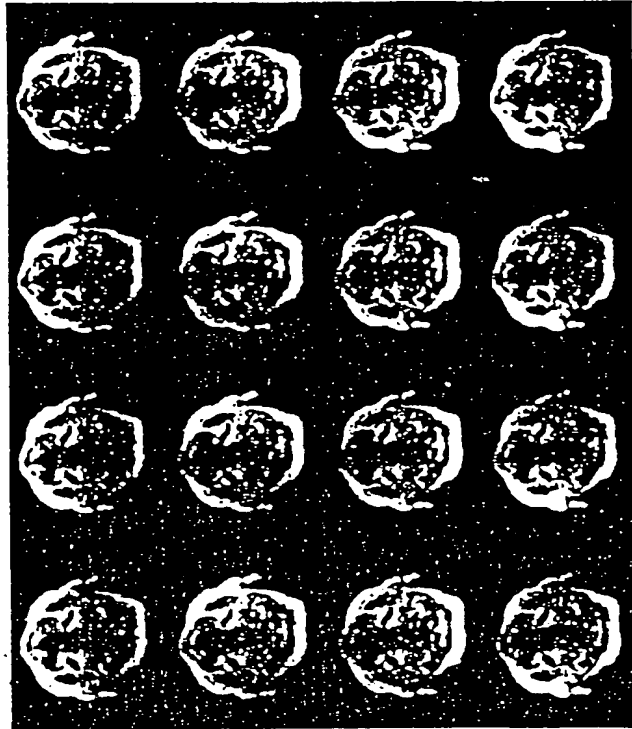


Figure 2: Simulation of the images received with a coil having the same sensitivity profile placed at 16 different positions around the brain.



**Figure 2:** Results of the reconstruction simulation using different sets of k-space lines. From left to right, the images were reconstructed by using the k-space lines acquired from skipping respectively 1, 5, 10 and 16 lines.

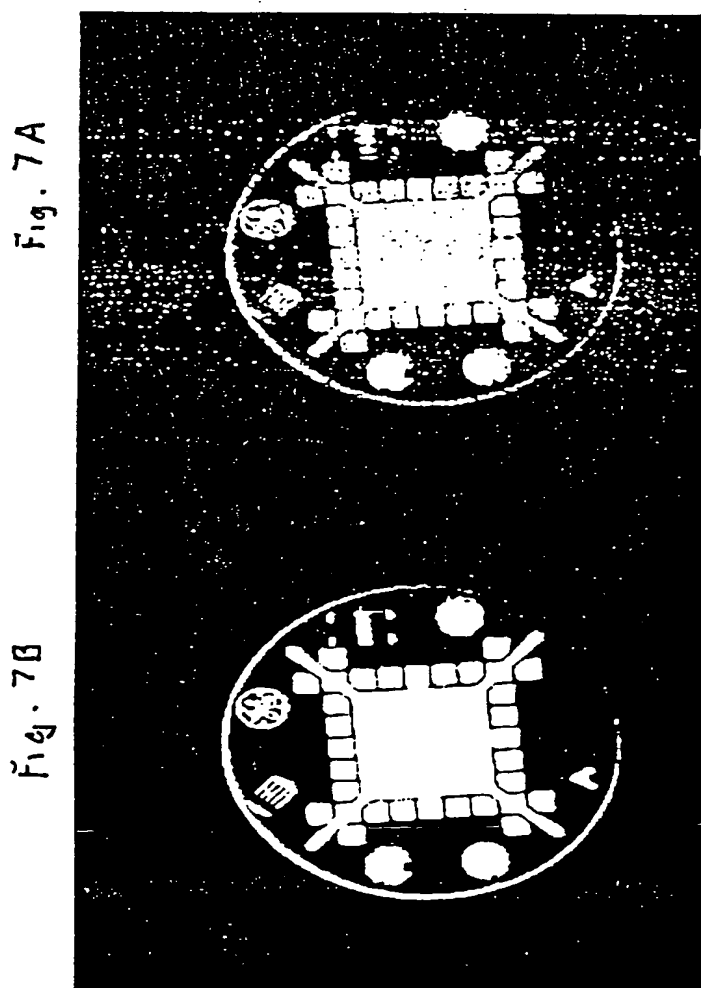


Figure 4- Experiment results: The left figure shows an image of a GE quality phantom acquired using a RARE sequence, and the image on the right shows the same image acquired with a 16 fold speed enhancement using our technique.

10/10

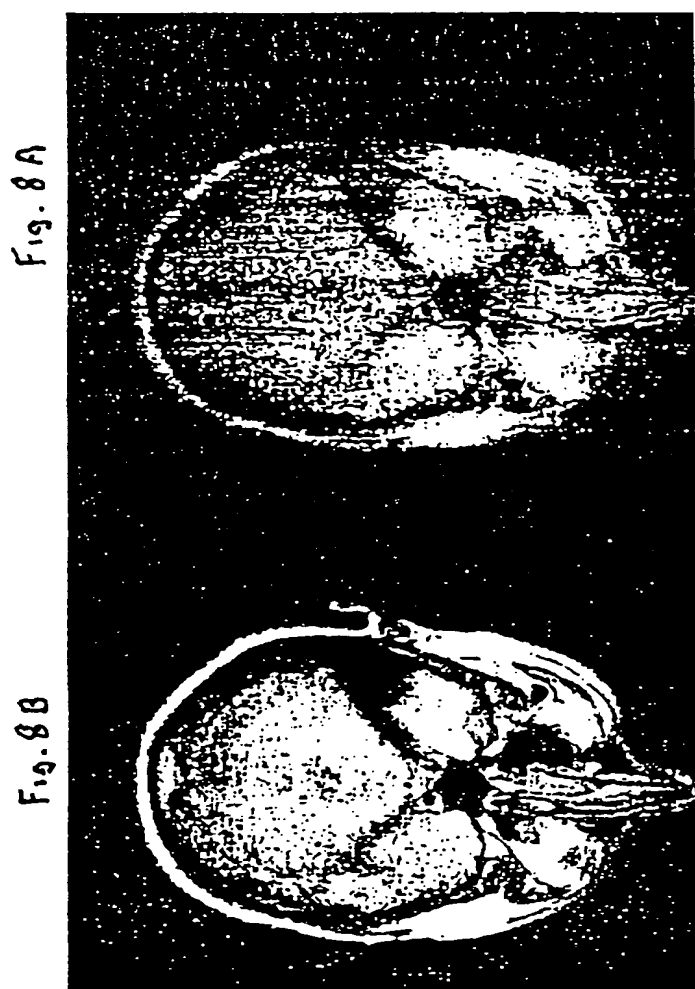


Figure-5: Experiment results: The left figure shows a brain image acquired using a RARE sequence, and the image on the right shows the same image acquired with a 16 fold speed enhancement using our technique.



# INTERNATIONAL SEARCH REPORT

International application No.  
PCT/US00/14171

## A. CLASSIFICATION OF SUBJECT MATTER

IPC(7) :G01V 3/00

US CL :324/307, 309

According to International Patent Classification (IPC) or to both national classification and IPC

## B. FIELDS SEARCHED

Minimum documentation searched (classification system followed by classification symbols)

U.S. : 324/307, 309, 314, 318, 322, 300

Documentation searched other than minimum documentation to the extent that such documents are included in the fields searched  
NONE

Electronic data base consulted during the international search (name of data base and, where practicable, search terms used)  
NONE

## C. DOCUMENTS CONSIDERED TO BE RELEVANT

Category*	Citation of document, with indication, where appropriate, of the relevant passages	Relevant to claim No.
X	US 5,587,656 A (VAN HEELS-BERGEN et al.) 24 December 1996 (24.12.1996), fig. 2, col. 1, col. 2, col. 5 lines 30-65.	1-5
A	US 5,712,567 A (WANG) 27 January 1998 (27.01.1998), fig. 4 and abstract.	1-5
A,P	US 5,945,826 A (LEUSSLER) 31 August 1999 (31.08.1999), abstract.	1-5

☐ Further documents are listed in the continuation of Box C.

☐ See patent family annex.

\* Special categories of cited documents:

"A" document defining the general state of the art which is not considered to be of particular relevance

"E" earlier document published on or after the international filing date

"L" document which may throw doubts on priority claim(s) or which is cited to establish the publication date of another citation or other special reason (as specified)

"O" document referring to an oral disclosure, use, exhibition or other means

"P" document published prior to the international filing date but later than the priority date claimed

"T"

later document published after the international filing date or priority date and not in conflict with the application but cited to understand the principle or theory underlying the invention

"X"

document of particular relevance; the claimed invention cannot be considered novel or cannot be considered to involve an inventive step when the document is taken alone

"Y"

document of particular relevance; the claimed invention cannot be considered to involve an inventive step when the document is combined with one or more other such documents, such combination being obvious to a person skilled in the art

"&"

document member of the same patent family

Date of the actual completion of the international search

29 JUNE 2000

Date of mailing of the international search report

19 JUL 2000

Name and mailing address of the ISA/US  
Commissioner of Patents and Trademarks  
Box PCT  
Washington, D.C. 20231

Facsimile No. (703) 305-3230

Authorized officer

LOUIS ARANA

Telephone No. (703) 305-4913

**THIS PAGE BLANK (USPTO)**

**This Page is Inserted by IFW Indexing and Scanning  
Operations and is not part of the Official Record**

**BEST AVAILABLE IMAGES**

Defective images within this document are accurate representations of the original documents submitted by the applicant.

Defects in the images include but are not limited to the items checked:

- ☐ BLACK BORDERS
- ☐ IMAGE CUT OFF AT TOP, BOTTOM OR SIDES
- ☐ FADED TEXT OR DRAWING
- ☐ BLURRED OR ILLEGIBLE TEXT OR DRAWING
- ☐ SKEWED/SLANTED IMAGES
- ☒ COLOR OR BLACK AND WHITE PHOTOGRAPHS
- ☐ GRAY SCALE DOCUMENTS
- ☐ LINES OR MARKS ON ORIGINAL DOCUMENT
- ☐ REFERENCE(S) OR EXHIBIT(S) SUBMITTED ARE POOR QUALITY
- ☐ OTHER: \_\_\_\_\_

**IMAGES ARE BEST AVAILABLE COPY.**

**As rescanning these documents will not correct the image problems checked, please do not report these problems to the IFW Image Problem Mailbox.**

**THIS PAGE BLANK (USPTO)**